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On the importance of using both T1-weighted and T2-weighted structural magnetic resonance imaging scans to model electric fields induced by non-invasive brain stimulation in SimNIBS

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Computational modeling of the electric fields (E-fields) induced by non-invasive brain stimulation has become increasingly popular and widely implemented in the past decade. E-field modeling is an informative tool that enables researchers to better understand the effects of non-invasive brain stimulation techniques such as transcranial magnetic stimulation (TMS) and transcranial direct current stimulation (tDCS) on the cortical and subcortical level. To date, E-field modeling studies have had widespread applications, such as informing how stroke damage alters cortical activation from TMS and tDCS [1], elucidating the impact of different tDCS montages on the magnitude and focality of stimulation [2,3], and unravelling dose-response relationships between tDCS-induced E-field strength and the degree of behavioral improvement [4]. Given the growing interest and diverse applications of E-field modeling, it is important to evaluate the factors that might affect the accuracy of E-field simulations, such as how to best construct anatomically valid head models.

A prerequisite for E-field modeling is the availability of a structural T1w magnetic resonance imaging (MRI) scan, which is segmented and meshed to create an anatomically valid head model. In addition to T1w scans, T2w scans are also recommended to improve the quality of the head mesh, making E-field modeling more accurate [5,6]. Nielsen et al. demonstrated that including both T1w + T2w scans results in consistently better tissue segmentation and decreased inter-individual mesh variance than T1w scans alone [5]. These findings were reproduced by Puonti et al., who reported that including a T2w scan improves segmentation accuracy in two commonly used E-field modeling software package: SimNIBS [7] and ROAST [8], and especially helps to make bone and cerebrospinal fluid segmentation more accurate [6]. Nevertheless, numerous studies perform E-field simulations on head models based on solely T1w scans. Thus, it is crucial to evaluate how the E-fields produced from T1w + T2w scans compare to those from T1w scans alone.

In this study, we aimed to build on the prior results in n = 30 participants in Nielsen et al. and Puonti et al.'s studies [5,6] by performing E-field modeling with T1w + T2w versus T1w scans alone on a larger sample size to capture a wider range of anatomical idiosyncrasies. In addition, we used the electrode placement of 4 × 1 high definition (HD-)tDCS montage to determine if these observed differences in T1w + T2w versus T1w alone still exist in a more focal montage than conventional primary motor cortex-supraorbital tDCS.

We evaluated the E-field strength difference between T1w + T2w versus T1w head meshing induced by HD-tDCS in 118 healthy participants (64 females, age range = 22–35 years), randomly selected from the Human Connectome Project (HCP) dataset [9]. In doing so, we aimed to provide insight into prior studies that have calculated the E-field magnitude solely based on T1w scans and guide future research considering to solely use T1w scans due to data availability and/or scanner limitations.

T1w and T2w MRI-scans were acquired with the Siemens MAGNETOM 3T scanner (for detailed scanning parameters, see Ref. [9]). Two finite element method tetrahedral head meshes were constructed per participant (Fig. 1A). The first mesh was constructed using both T1w + T2w scans. In contrast, the second mesh was based only on a T1w scan. Head model reconstruction was performed using the headreco command [5], which automatically segments and meshes MRI scans.

We modelled 4 × 1 HD-tDCS in SimNIBS (v3.2.3) [7] in each participant's T1w + T2w and T1w head mesh, for a total of 236 paired E-field models (118 participants x 2 models per person). In each model, a circular anode was placed over C3 (primary motor cortex) and four circular cathodes over FC3, C1, CP3 and C5 (all 0.25 cm electrode radius). The modelled stimulation intensity was 1 mA over the C3 anode and 0.25 mA at each cathode. We used standard conductivity values for the modelled tissues (white matter: 0.126 S/m, grey matter: 0.275 S/m, cerebrospinal fluid: 1.654 S/m, bone: 0.01 S/m, skin: 0.465 S/m, and eyes: 0.5 S/m). For both the T1w + T2w and the T1w mesh simulations, we extracted the average E-field induced in the primary motor cortex using a region

Abbreviations: E-field, electric Field; HD, high-definition; MRI, magnetic resonance imaging; tDCS, transcranial direct current stimulation; TMS, transcranial magnetic stimulation.

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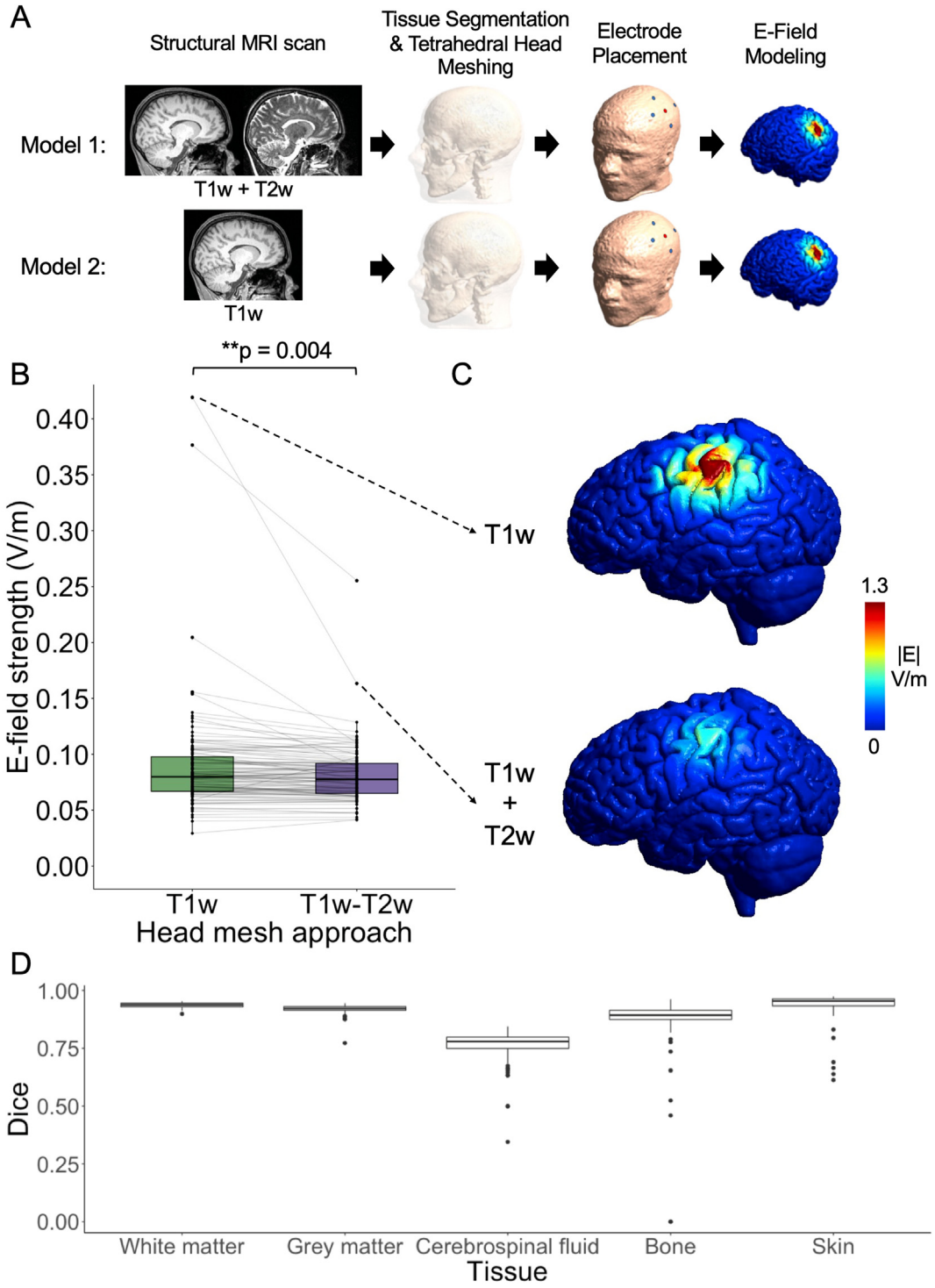

Fig. 1. Difference in electric fields induced in head models based on T1w and T2w MRI scans. **A.** E-field modeling pipeline for the T1w and T1w + T2w meshes. We constructed two head meshes per participant: one mesh from the T1w scan alone and one from the T1w + T2w scans. **B.** Boxplot demonstrating E-field strength in primary motor cortex induced by 4 × 1 high definition transcranial electrical stimulation (HD-tDCS) in mesh based on a T1w scan and T1w + T2w scan. **C.** Visualization of E-fields induced in the participant with the highest E-field as a result of tES in T1w + T2w versus T1w head meshes. **D.** Sørensen–Dice scores of T1w + T2w meshing versus T1w meshing. A score of 1 indicates a high overlap between both meshing procedures for the tissue whereas a score of 0 indicates no overlap between both procedures. These data suggest that the integrity of bone and cerebrospinal fluid segmentations are most affected by T1w only meshes.

of interest (ROI) analysis. We centered the ROI at the subject space transformed peak MNI coordinate of the primary motor cortex (x = −37, y = −21, z = 58) and extracted the average E-field in a 10 mm radius sphere in each model using a grey matter mask [10].

E-fields induced in head models constructed solely by T1w scans were significantly different from E-fields induced in head models constructed from T1w and T2w MRI scans, as determined by a paired T-test in RStudio ($t(117) = 2.9484$, $p = 0.004$) [11] (Fig. 1B). Moreover, upon removal of the outlier values, defined as the mean ± 2 * standard deviation, the significant difference remained present ($t(114) = 4.0897$, $p > 0.001$). The group-average E-field in the primary motor cortex in models created by T1w + T2w meshing was 0.082 ± 0.026 V/m (mean ± standard deviation, variance coefficient: 31.69%). In comparison, the group-average E-field induced in the same subjects using solely T1w scans for meshing was 0.089 ± 0.048 (variance coefficient: 53.81%). The difference between both procedures was starkest in the head models with the highest E-fields (Fig. 1C).

To elucidate what caused these E-field magnitude differences between the T1w + T2w versus T1w only meshes, we calculated the Sørensen–Dice index comparing the within-subject differences in tissue masks per tissue type (Fig. 1D). A Sørensen–Dice index value of 1 indicates perfect overlap between T1w and T1w + T2w tissue masks whereas a value of 0 represents no overlap. The following mean Sørensen–Dice values were obtained: white matter: 0.92 ± 0.02, grey matter: 0.94 ± 0.01, cerebrospinal fluid: 0.76 ± 0.07, bone: 0.87 ± 0.13, skin: 0.94 ± 0.06. Consistent with previous work [5,6], the two meshing approaches of T1w + T2w versus T1w were most divergent in cerebrospinal fluid and bone segmentations. Qualitative inspection of the meshes and previous literature [5,6] indicate that T1w + T2w scanning resulted in more accurate segmentation of the bone − cerebrospinal fluid border. For instance, in two participants, the bone dice index was 0. In these two cases, the T1w procedure had incorrectly selected cerebrospinal fluid as bone tissue while the T1w + T2w procedure correctly segmented bone.

Taken together, including T1w + T2w MRI scans in the meshing approach diminished interindividual E-field variance, mainly due to better bone and cerebrospinal fluid tissue segmentation. These data suggest that less accurate tissue segmentations, particularly in the meshes created from only T1w scans, are a meaningful source of E-field variance that can affect the fidelity and interpretation of models. Our findings, combined with the results of Nielsen et al. and Puonti et al., emphasize the importance of using both T1w and T2w scans for E-field modeling [5,6]. While we only used SimNIBS, the work of Puonti et al. might help translate our results to ROAST, as they reported that both modeling software packages benefit similarly from an additional T2w-scan.

These E-field modeling data add to the literature by including a larger dataset of 118 participants that captures more interindividual anatomical idiosyncrasies and quantifies how HD-tDCS induced E-fields are impacted by T1w + T2w scan meshes compared to meshes made from only T1w scans. In summary, computational E-field strength is significantly impacted by the inclusion of both T1w + T2w MRI scans. Whenever possible, future E-field modeling studies should consider the inclusion of both T1w + T2w structural MRI scans for more accurate E-field modeling and a better representation of cerebrospinal fluid and bone tissue. Although we used SimNIBS modeling, there is theoretical ground to assume that these results are also valid for ROAST modeling, particularly as both software packages utilize similar analyses procedures such as using SPM12 for tissue segmentation processes.

Conflict of interest statement

We confirm that there are no known conflicts of interest associated with this publication and there was no financial support for this work that could have influenced its outcome.

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